Fithogonal mag.

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☐ 1. Document ID: US 5492123 A Relevance Rank: 65

L5: Entry 3 of 4 File: USPT Feb 20, 1996

US-PAT-NO: 5492123
DOCUMENT-IDENTIFIER: US 5492123 A

TITLE: Diffusion weighted magnetic resonance imaging

DATE-ISSUED: February 20, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Edelman; Robert R.	Wellesley	MA		

ASSIGNEE-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY	TYPE CODE
Siemens Medical Systems, Inc.	Iselin	NJ			02
Beth Isreal Hospital	Boston	MA			02

APPL-NO: 8/ 106228
DATE FILED: August 13, 1993

INT-CL: [6] A61B 5/055
US-CL-ISSUED: 128/653.2; 128/653.4, 128/708
US-CL-CURRENT: 600/410; 600/413, 600/420, 600/521
FIELD-OF-SEARCH: 128/653.2, 128/653.3, 128/653.4, 128/708, 324/307, 324/309, 324/306

PRIOR-ART-DISCLOSED:

U.S. PATENT DOCUMENTS

PAT-NO	ISSUE-DATE	PATENTEE-NAME	US-CL
5079503	January 1992	Siebold et al.	324/309
5084675	January 1992	Erich et al.	324/309
5162730	November 1992	Schmitt et al.	324/309

FOREIGN PATENT DOCUMENTS

FOREIGN-PAT-NO	PUBN-DATE	COUNTRY	US-CL
0076054	September 1982	EPX	
429715A1	December 1989	EPX	

OTHER PUBLICATIONS

Koresec et al., "Ultra-Fast MR Angiography Using A Velocity-Selective Prep Pulse and many Readouts" p. 214.

Frahm et al., "High-Speed STEM MRI of the Human Heart" Magnetic Resonance in Medicine 22:133-142 (1991).
Zhang et al., "Measurement of Rat Brian Perfusion by NMR Using Spin Labeling of Arterial Water: In Vivo Determination of the Degree of Spin Labeling" MRM 29:416-421 (1993).
Moseley et al., "Diffusion-weighted MR Imaging of Anisotropic Water Diffusion in Cat Central Nervous System" Radiology, 176:439-445 (1990).
Edelman, "MR Angiography: Present and Future", AJR, 161:1-11 (1993).
Le Bihan, "Molecular Diffusion Nuclear magnetic Resonance Imaging" Magnetic Resonance Quarterly, 7(1):1-30 (1991).
Beaulieu et al., "Diffusion-Weighted MR Microscopy with Fast Spin-Echo" MRM, 30:201-206 (1993).
Warach et al., "Fast Magnetic Resonance Diffusion-Weighted Imaging of Acute Human Stroke" Neurology, 42:1717-1723 (1992).
Morvan et al., "Simultaneous Temperature and Regional Blood Volume Measurements in Human Muscle Using an MRI Fast Diffusion Technique" MRM, 29:371-377 (1993).
Stehling et al., "Whole-Body Echo-Planar MR Imaging at 0.5T.sup.1 ", Radiology, 170:257-263 (1989).
Moseley et al., "Diffusion-Weighted MR Imaging of Acute Stroke: Correlation with T2-Weighted and magnetic Susceptibility-Enhanced MR Imaging in Cats" Amer. J. Neuralradiology, 11:423-429 (May/Jun. 1990).

ART-UNIT: 335
PRIMARY-EXAMINER: Smith; Ruth S.
ATTY-AGENT-FIRM: Jay; Mark H.

ABSTRACT:

The present invention relates to the use of diffusion weighted magnetic resonance imaging procedures for the diagnosis of conditions in which blood circulation or similar molecular displacements within biological tissues are measured. Pulse sequences are used in accordance with the invention which suppress the contribution of non-moving tissue or bulk motion.

14 Claims, 4 Drawing figures

Full | Title | Citation | Front | Review | Classification | Date | Reference   KWIC | Draw Desc | Image

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☐ 2. Document ID: US 5699801 A Relevance Rank: 65

L5: Entry 2 of 4 File: USPT Dec 23, 1997

US-PAT-NO: 5699801
DOCUMENT-IDENTIFIER: US 5699801 A

TITLE: Method of internal magnetic resonance imaging and spectroscopic analysis and associated apparatus

DATE-ISSUED: December 23, 1997

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Atalar; Ergin	Columbia	MD		
Bottomley; Paul A.	Columbia	MD		
Zerhouni; Elias A.	Baltimore	MD		

ASSIGNEE-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY	TYPE CODE
The Johns Hopkins University	Baltimore	MD			02

APPL-NO: 8/ 457833
DATE FILED: June 1, 1995

INT-CL: [6] A61B 5/055
US-CL-ISSUED: 128/653.2; 128/653.5, 324/318, 324/322
US-CL-CURRENT: 600/410; 324/318, 324/322, 600/422
FIELD-OF-SEARCH: 128/653.2, 128/653.5, 128/653.1, 324/307, 324/309, 324/318, 324/322

PRIOR-ART-DISCLOSED:

U.S. PATENT DOCUMENTS

PAT-NO	ISSUE-DATE	PATENTEE-NAME	US-CL
4672972	June 1987	Berke	128/653
4766381	August 1988	Conturo et al.	324/309
4932411	June 1990	Fritschy et al.	128/653A
5170789	December 1992	Narayan et al.	128/653.5
5271400	December 1993	Dumoulin et al.	128/653.2
5293872	March 1994	Alfano et al.	128/664
5307808	May 1994	Dumoulin et al.	128/653.2
5473251	December 1995	Mori	128/653.5

FOREIGN PATENT DOCUMENTS

FOREIGN-PAT-NO	PUBN-DATE	COUNTRY	US-CL
0469035	February 1994	EPX	

OTHER PUBLICATIONS

Kantor et al., "In vivo P Nuclear Magnetic Resonance Measurements in Cacine Heart Using a Catheter-Coil," Circulation Research, vol. 55, pp. 261, 266 (Aug. 1984).
Edelstein et al., "Electronic Decoupling of Surface-Coil Receivers for NMR Imaging and Spectroscopy," Journal of Magnetic Resonance, vol. 67, pp. 156-161 (1986).
Merickel et al., "Identification and 3-d Quantification of Atherosclerosis Using Magnetic Resonance Imaging," Comput. Biol. Med., vol. 18, pp. 89-102 (1988).
Martin et al., "Inflatable Surface Coil for MR Imaging of the Prostate", Radiology, Apr. 1988, pp. 260-270.
Maynor et al., "Chemical Shift Imaging of Atherosclerosis at 7.0 Tesla," Investigative Radiology, vol. 24, pp. 52-60 (1989).
Mohiaddin et al., "Chemical Shift Magnetic Resonance Imaging of Human Atheroma," Br. Heart J., vol. 62, pp. 81-89 (1989).
Asdente et al., "Evaluation of Atherosclerotic Lesions Using NMR Microimaging," Atherosclerosis, vol. 80, pp. 243-253 (1990).
Pearlman et al., "Nuclear Magnetic Resonance Microscopy of Atheroma in Human Coronary Arteries," Angiology, vol. 42, pp. 726-733 (1991).
Vinitski et al., "Magnetic Resonance Chemical Shift Imaging and Spectroscopy of Atherosclerotic Plaque," Investigative Radiology, vol. 26, pp. 703-714 (1991).
Waller et al., "Intravascular Ultrasound: A Histological Study of Vessel During Life," Circulation, vol. 85, pp. 2305-2310 (1992).
Martin et al., "MR Imaging of Blood Vessel with an Intravascular Coil," J. Magn. Reson. Imaging, vol. 2, pp. 421-429 (1992).
Hurst et al., "Intravascular (Catheter) NMR Receiver Probe: Preliminary Design Analysis and Application to Canine Iliofemoral Imaging," Magn. Reson, Med., vol. 24, pp. 343-357 (Apr. 1992).
Dumoulin et al., "Real-time Position Monitoring of Invasive Devices Using Magnetic Resonance," Magnetic Resonance in Medicine, vol. 29, pp. 411-415 (Mar. 1993).
Spears et al., "In Vivo Coronary Angioscopy," Journal of the American College of Cardiology, vol. 1, No. 5, pp. 1311-1314 (1983).
Koechli et al., "Catheters and Guide Wires for Use in an Echo-Planar MR Fluoroscopy System," R. 79th Scientific Meeting, editor, Radiology, vol. 189 (P), p. 319 (Nov. 1993).
Abstract, McDonald et al., "Performance Comparison of Several Coil Geometries for

Use in Catheters," R. 79th Scientific Meeting, editor, Radiology, vol. 189(P) p. 319 (Nov. 1993).
Merickel et al., "Noninvasive Quantitative Evaluation of Atherosclerosis Using MRI and Image Analysis," Arteriosclerosis and Thrombosis, vol. 13, pp. 1180-1186 (1993).
Yuan et al., "Techniques for High-Resolution MR Imaging of Atherosclerotic Plaques," J. Magnetic Resonance Imaging, vol. 4, pp. 43-49 (1994).
Martin et al., "Intravascular MR Imaging in a Porcine Animal Model," Magn. Reson, Med. vol. 32, pp. 224-229 (Aug. 1994).

ART-UNIT: 335
PRIMARY-EXAMINER: Smith; Ruth S.
ATTY-AGENT-FIRM: Silverman; Arnold B. Eckert Seamans Cherin & Mellott, LLC

ABSTRACT:

The invention provides a method for magnetic resonance imaging and spectroscopic analysis of the interior of a specimen which includes positioning the specimen within a main magnetic field, introducing an invasive probe having an elongated receiver coil into or adjacent to the specimen with the coil having at least one pair of elongated electrical conductors, preferably, generally parallel to each other disposed within a dielectric material and having a pair of ends electrically connected to each other. RF pulses are provided to the region of interest to excite magnetic resonance signals, gradient magnetic pulses are applied to the region of interest with the receiver coil receiving magnetic resonance signals and emitting responsive output signals which may be processed by a computer to provide image information for display in a desired manner. The method in a preferred form involves employing a flexible receiver coil which has uniform sensitivity along the coil and may be operated even when the magnetic resonance signal is in an oblique position. Tuning capacitance may be distributed along the length of the coil and/or a Faraday screen provided to minimize dielectric losses between the coil and the surrounding material of the specimen. The method may be used on a wide variety of specimens and in a preferred use is introduced into small blood vessels of a patient to facilitate determination of atherosclerotic plaque. Medical intervention procedures, such as plaque removal, may be employed generally simultaneously with the imaging of the present invention. Corresponding apparatus is provided.

85 Claims, 32 Drawing figures

Full | Title | Citation | Front | Review | Classification | Date | Reference   KWIC | Draw Desc | Image

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☐ 3. Document ID: US 4684891 A   Relevance Rank: 58

L5: Entry 4 of 4   File: USPT   Aug 4, 1987

US-PAT-NO: 4684891
DOCUMENT-IDENTIFIER: US 4684891 A

TITLE: Rapid magnetic resonance imaging using multiple phase encoded spin echoes in each of plural measurement cycles

DATE-ISSUED: August 4, 1987

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Feinberg; David A.	Berkeley	CA		

ASSIGNEE-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY	TYPE CODE
The Regents of the University of California	Berkeley	CA			02

APPL-NO: 6/ 760832
DATE FILED: July 31, 1985

INT-CL: [4] G01R 33/20
US-CL-ISSUED: 324/309; 324/307
US-CL-CURRENT: 324/309; 324/307
FIELD-OF-SEARCH: 324/300, 324/307, 324/309, 324/310, 324/311, 324/312, 324/318, 324/322

PRIOR-ART-DISCLOSED:

U.S. PATENT DOCUMENTS

PAT-NO	ISSUE-DATE	PATENTEE-NAME	US-CL
4070611	January 1978	Earnst	324/307
4297637	October 1981	Crooks et al.	324/309
4318043	March 1982	Crooks et al.	324/309
4355282	October 1982	Young	324/309
4471305	September 1984	Crooks et al.	324/309
4567440	January 1986	Haselgrove	324/309
4570119	February 1986	Wehrli	324/309
4587489	May 1986	Wehrli	324/309

OTHER PUBLICATIONS

Kumar et al, J. Mag., Res. 18, 69-83 (1975).
Mansfield et al, J. Mag., Res. 29. 335-373 (1978).

ART-UNIT: 265
PRIMARY-EXAMINER: Tokar; Michael J.
ATTY-AGENT-FIRM: Nixon & Vanderhye

ABSTRACT:

Slice selective 90.degree. and plural subsequent 180.degree. NMR RF pulses are utilized to elicit a train of NMR spin echoes from a given slice or "planar volume" of the object under test in each of plural measurement cycles. Spatial information is encoded within the spin echo by imposing a G.sub.x gradient during each spin echo readout. Phase encoding in a second G.sub.y dimension is achieved by using (1) a cycle-dependent .beta.G.sub.y gradient at least once during each NMR measurement cycle and (2) further .delta.G.sub.y magnetic gradient pulses in association with some or all of the individual spin echo responses within each measurement cycle. The two different types of G.sub.y gradient pulses are dimensioned and timed so as to result in the desired number of phase encoded spin echo signals which subsequently can be arranged in a linearly increasing progression of phase encoding so as to be usable in a two-dimensional Fourier transformation process to produce an NMR image. Because the spin echo data are taken at different times of echo occurrences within a given measurement cycle, T2 artifact may be present in such an image. However T2 correction may be provided by calculating T2 and scaling all of the time domain spin echo data to a single common equivalent time of echo occurrence before performing the final two-dimensional Fourier transformation process which results in a final T2-corrected NMR image.

26 Claims, 12 Drawing figures

Full | Title | Citation | Front | Review | Classification | Date | Reference   KWIC | Draw Desc | Image

☐ 4. Document ID: US 5898306 A Relevance Rank: 56

L5: Entry 1 of 4 File: USPT Apr 27, 1999

US-PAT-NO: 5898306
DOCUMENT-IDENTIFIER: US 5898306 A

TITLE: Single circuit ladder resonator quadrature surface RF coil

DATE-ISSUED: April 27, 1999

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Liu; Haiying	Minneapolis	MN		
Truwit; Charles L.	Wayzata	MN		

ASSIGNEE-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY	TYPE CODE
Regents of the University of Minnesota	Minneapolis	MN			02

APPL-NO: 8/ 838604
DATE FILED: April 9, 1997

INT-CL: [6] G01V 3/00
US-CL-ISSUED: 324/322; 324/318
US-CL-CURRENT: 324/322; 324/318
FIELD-OF-SEARCH: 324/318, 324/322, 324/300, 324/314, 324/312, 324/307, 324/309

PRIOR-ART-DISCLOSED:

U.S. PATENT DOCUMENTS

PAT-NO	ISSUE-DATE	PATENTEE-NAME	US-CL
4707664	November 1987	Fehn et al.	324/322
4721913	January 1988	Hyde et al.	324/318
4752738	June 1988	Patrick et al.	324/318
4816765	March 1989	Boskamp	324/318
4839594	June 1989	Misic et al.	324/318
4879516	November 1989	Mehdizadeh et al.	324/318
4881032	November 1989	Bottomley et al.	324/309
4906933	March 1990	Keren	324/318
4918388	April 1990	Mehdizadeh et al.	324/322
4931734	June 1990	Kemner et al.	324/318
4985678	January 1991	Gangarosa et al.	324/318
5030915	July 1991	Boskamp et al.	324/318
5045792	September 1991	Mehdizadeh	324/318
5144240	September 1992	Mehdizadeh et al.	324/318
5160891	November 1992	Keren	324/318
5196796	March 1993	Misic et al.	324/322
5212450	May 1993	Murphy-Boesch et al.	324/322
5235277	August 1993	Wichern	324/318
5280248	January 1994	Zou et al.	324/318
5285160	February 1994	Loos et al.	324/318
5365173	November 1994	Zou et al.	324/322
5374890	December 1994	Zou et al.	324/318
5394087	February 1995	Molyneaux	324/318
5430378	July 1995	Jones	324/318
5521506	May 1996	Misic et al.	324/322

OTHER PUBLICATIONS

Ballon, D., et al., "A 64 MHz Half-Birdcage Resonator for Clinical Imaging", J. of Magnetic Resonance, 90, 131-140, (1990).
Hu, X., et al., "Reduction of Field of View for Dynamic Imaging", Magnetic Resonance in Medicine, 31, No. 6, 691-694, (1994).
Mehdizadeh, M., "RF Coils for Magnetic Resonance Imaging", RF Design, 29-38, (1991).
Panych, L.P., et al., "A Dynamically Adaptive Imaging Algorithm for Wavelet-Encoded MRI", Magnetic Resonance in Medicine, 32, No. 6, 738-746, (1994).

ART-UNIT: 287
PRIMARY-EXAMINER: Arana; Louis
ATTY-AGENT-FIRM: Schwegman, Lundberg, Woessner, and Kluth, P.A.

ABSTRACT:

A single-circuit quadrature surface coil is formed from two ladder resonator coils and includes a first mode circuit path for detecting or generating magnetic flux in a vertical axis from a body under investigation and a second mode circuit path for detecting or generating magnetic flux in a parallel axis, with the first mode and second mode currents 90 degrees out of phase. The surface coil, which supports two resonance current modes for quadrature operation on only one single coil conductor structure, provides a high signal-to-noise ratio (SNR) and a good B.sub.1 homogeneity over the imaging volume. This coil alone may be used either for both transmitting and receiving RF signals or for detecting RF signals as "receive only." This coil is well suited for imaging the human neck, spine and heart.

17 Claims, 7 Drawing figures

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Term	Documents
ORTHOGONAL.USPT.	79010
ORTHOGONALS.USPT.	20
MAGNETIC.USPT.	327513
MAGNETICS.USPT.	7243
GRADIENT.USPT.	105989
GRADIENTS.USPT.	30464
(3 AND (ORTHOGONAL ADJ MAGNETIC) AND GRADIENT).USPT.	4

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L5: Entry 4 of 4 File: USPT Aug 4, 1987

DOCUMENT-IDENTIFIER: US 4684891 A
TITLE: Rapid magnetic resonance imaging using multiple phase encoded spin echoes in each of plural measurement cycles

ABPL:
Slice selective 90.degree. and plural subsequent 180.degree. NMR RF pulses are utilized to elicit a train of NMR spin echoes from a given slice or "planar volume" of the object under test in each of plural measurement cycles. Spatial information is encoded within the spin echo by imposing a G.sub.x gradient during each spin echo readout. Phase encoding in a second G.sub.y dimension is achieved by using (1) a cycle-dependent .beta.G.sub.y gradient at least once during each NMR measurement cycle and (2) further .delta.G.sub.y magnetic gradient pulses in association with some or all of the individual spin echo responses within each measurement cycle. The two different types of G.sub.y gradient pulses are dimensioned and timed so as to result in the desired number of phase encoded spin echo signals which subsequently can be arranged in a linearly increasing progression of phase encoding so as to be usable in a two-dimensional Fourier transformation process to produce an NMR image. Because the spin echo data are taken at different times of echo occurrences within a given measurement cycle, T2 artifact may be present in such an image. However T2 correction may be provided by calculating T2 and scaling all of the time domain spin echo data to a single common equivalent time of echo occurrence before performing the final two-dimensional Fourier transformation process which results in a final T2-corrected NMR image.

BSPR:
This invention is related to the field of mangetic resonance imaging (MRI) utilizing nuclear magnetic resonance (NMR) phenomena. It is particularly related to novel apparatus and method for acquisition of NMR image data (and for T2-dependence correction of same). It uses an NMR imaging technique wherein slice-selective NMR RF excitations including a train of 180.degree. nutation pulses are utilized to elicit a corresponding train of plural NMR spin echo RF signal responses (which are readout during an imposed magnetic gradient pulse along a first dimension). This invention provides a novel and advantageous technique for more rapidly acquiring such NMR spin echo RF signals with requisite phase encoding in a second dimension where relatively large "whole body" sizes are involved.

BSPR:
Magnetic resonance imaging (MRI) is now coming into widespread commercial usage. Nevertheless, there are still many possible areas of improvement. One such area for potential improvement relates to the time required to acquire the NMR image data for a complete image in a "whole body" type of scanner (i.e. dimensioned to accept an entire human body within a cryogenic static magnetic field coil as well as included magnetic gradient coils and RF signal transmission/reception coils). The necessary relatively large dimension of the gradient coils causes inherent inductance which, in turn, limits the speed with which the various magnetic gradient coils can be effectively switched on and off during a given data acquisition cycle.

BSPR:
Prior MRI data acquisition techniques can be grouped into at least three different categories distinguished by the way in which spatial encoding is achieved using magnetic gradients.

BSPR:

Multiple section imaging as performed by Crooks et al (see the above referenced related U.S. patents and patent applications and see also Kumar, Welti, Earnst at J. Mag. Res. 18, 69-83, 1975) utilizes slice selective magnetic gradient pulses which are "on" during each radio frequency pulse (e.g. 90.degree. nutation pulses and 180.degree. nutation pulses) so as to achieve NMR at the Larmor frequency within a selected planar volume without substantially disturbing the spin lattice of adjacent planar volumes (each incident radio frequency pulse is typically modulated by a sinc function so as to select a substantially square edged planar volume in the spatial domain). After extracting the useful spin echo NMR RF response from a given planar volume, it is allowed to relax to its quiescent alignment with a static z-axis magnetic field while, in the meantime, other planar volumes are similarly selectively defined by suitable magnetic gradient pulses and sinc-modulated RF NMR pulses so as to produce the desired NMR spin echo responses from these other planar volumes. After a sequence of planar volumes have thus been irradiated and their respective NMR responses captured for subsequent analysis, the entire sequence is repeated many times with incrementally increased magnetic gradient along an orthogonal y-axis so as to encode spatial information. Spatial information for the second x-axis dimension is encoded by imposing a constant magnetic gradient pulse along the x-axis during each spin echo NMR signal readout. The y-axis phase encoding is changed for each of M NMR cycles so as to provide a linearly increasing progression of y-axis phase encoding (the number of resulting image lines along the y-axis will be equal to the number M of phase encoding cycles of the sequence). A two-dimensional Fourier transformation process is then utilized to obtain the final NMR image (see above referenced U.S. Pat. No. 4,599,565).

BSPR:
A very fast data acquisition technique has been proposed by Mansfield et al in J. Mag. Res. 29, p. 335-373 (1978) which is sometimes called "Echo Planar Imaging". Here, all of the required NMR image data is obtained in a single NMR pulse sequence thus avoiding the need to let a given volume "relax" between repeated cycles of NMR pulse sequences. Relatively rapid modulation of an NMR RF response signal is achieved using a rapidly switched magnetic gradient so as to encode spatial information within a single train of NMR response signals. Indeed, two dimensions of spatial information can be encoded within a single NMR measurement cycle sequence. Reportedly less than 100 milliseconds per image is required for data acquisition using this Mansfield technique.

BSPR:
However, because the Mansfield technique requires very rapid switching of magnetic field gradient pulses (e.g. so as to produce the requisite large number of sustained spin echoes) it cannot be used with the large size magnetic gradient coils required for "whole body" MRI scanners. Rather, only the use of relatively small gradient coils (with reduced inherent inductance) and higher applied voltage/current levels can achieve the relatively fast magnetic gradient rise times (e.g. in terms of microseconds) required for practical implementation of the Mansfield et al technique. Young (U.S. Pat. No. 4,355,282) proposes a modification to this echo planar imaging technique wherein pulsed magnetic gradients increase in phase encoding amplitude for successive FID signals and on successive cycles of the pulse sequence.

BSPR:
Yet another group of MRI data acquisition techniques (sometimes termed "three-dimensional volume imaging", see Earnst, U.S. Pat. No. 4,070,611) do not use slice selective magnetic gradients to define multiple image volumes. Instead, a three-dimensional Fourier transformation uses phase encoding magnetic gradients on each spatial axis perpendicular to each image plane. Two orthogonal magnetic gradient axes produce cycle dependent changes in phase for M image lines and S number of image slices or sections. The total number of pulse sequence cycles is then M.times.S.times.C, where C is the number of cycles or repetitions of the sequence.

BSPR:
Currently, most clinically useful MRI scanners utilize Fourier transformation reconstruction techniques and require data acquisition times typically between 5 and 40 minutes. However, the present invention can achieve clinically useful MRI with significantly reduced data acquisition times (e.g. between 5 and 40 seconds) achieved using relatively slower magnetic gradient switching times (e.g. in the millisecond range) that can be practically realized with whole body sized MRI

scanners.

BSPR:
Of the prior MRI data acquisition techniques, the Crooks et al technique has many advantages which are retained by the improvement of this invention. For example, the Crooks et al technique of using 180.degree. RF nutation pulses to produce spin echo RF responses (instead of utilizing rapidly switched magnetic gradients to elicit spin echoes) results in substantial cancellation of signal artifacts at each spin echo time of occurrence otherwise expected due to background magnetic field inhomogeneities. Such static magnetic field inhomogeneities result in accumulated errors in the echo planar technique or in switched magnetic gradient techniques used for signal refocusing and thus cause errors in the spatially-encoded phase information. Furthermore, using the Crooks et al 180.degree. RF nutation pulses to elicit multiple spin echoes avoids accumulation of signal artifact caused by NMR responses due to the chemical shift of nuclei (e.g. large apparent spatial separation of water and fat nuclei signals is thus avoided). On the other hand, echo planar or magnetic gradient reversal techniques for refocusing the FID cause phase information resulting from chemical shift to accumulate in each successive spin echo response. Furthermore, the Crooks et al technique of using 180.degree. RF nutation pulses to produce spin echoes permits much longer spin echo signal formation times to be achieved thus producing higher signal to noise ratios.

BSPR:
The Crooks et al technique may produce a "ghost" artifact in the image. However, I have discovered that such an artifact may be reduced or eliminated by using slice selections magnetic gradients pulses of alternating polarity.

BSPR:
In this invention, y-axis spatial information is phase encoded within a multiple spin echo pulse sequence using two different types and/or increments of phase change. For example, a cycle-dependent y-axis .beta. magnetic gradient pulse is employed at least once during each basic NMR pulse sequence. In addition, incremental y-axis .delta. magnetic gradient pulses are employed between some or all of the individual spin echoes within a given NMR measurement cycle sequence so as to produce a succession of y-axis phase encoded spin echoes during a single NMR pulse measurement sequence or cycle. Since the magnitude of spin echo responses exponentially decreases in accordance with the T2 NMR parameter, it is not possible to obtain all of the desired y-axis phase encoded spin echo data within a single measurement cycle. Accordingly, plural measurement cycles are still required and the cycle-dependent y-axis .beta. gradient pulse is utilized in conjunction with the incremental y-axis .delta. gradient pulses so as to produce a requisite number of y-axis phase encoded spin echoes (i.e. with a linear progression of encoded phase increment between each of successive spin echoes--perhaps after a re-organizational grouping during a data processing phase) after a relatively fewer number of repeated measurement cycles than was heretofore required using the Crooks et al technique. Accordingly, the overall data acquisition time required for a given image has been reduced.

DRPR:
FIG. 2B depicts an improved G.sub.z gradient pulse sequence which may be used to reduce or eliminate the artifact depicted in FIG. 2A;

DEPR:
Typically, a human or animal subject (or other object) 10 is inserted along the z-axis of a static cryogenic magnet which establishes a substantially uniform magnetic field directed along the z-axis within the portion of the object of interest. Gradients may be imposed within this z-axis directed magnetic field along the x, y or z axes by a set of x, y, z gradient amplifiers and coils 14. NMR RF signals are transmitted into the body 10 and NMR RF responses are received from the body 10 via RF coils 16 connected by a conventional transmit/receive switch 18 to an RF transmitter 20 and RF receiver 22.

DEPR:
As will be apparent to those in the art, such an arrangement may be utilized so as to generate desired sequences of magnetic gradient pulses and NMR RF pulses and to measure desired NMR RF responses in accordance with stored computer programs. As depicted in FIG. 1, the MRI system of this invention will typically include RAM, ROM and/or other stored program media adapted (in accordance with

the following descriptions) so as to generate multiple phase encoded spin echoes during each of multiple measurement cycles and to process the resulting MRI data into a final high resolution NMR image.

DEPR:
FIG. 2 depicts a typical prior art Crooks et al data acquisition sequence. For example, each measurement cycle may be initiated by a 90.degree. NMR RF excitation followed by a 180.degree. NMR RF nutation pulse located .tau. later in time and, if desired for signal averaging purposes, followed by subsequent 180.degree. RF nutation pulses (typically distributed at 2.tau. time intervals). It will be noted that during each RF excitation pulse there is a slice selection G.sub.z magnetic gradient pulse switched "on" so as to selectively excite only the desired "slice" or "planar volume" (e.g. a slice of given relatively small thickness through the object being imaged). During each resulting spin echo NMR RF response, x-axis phase encoding is achieved by applying an x-axis magnetic gradient during the readout procedure (typically each spin echo pulse is sampled every 30 microseconds or so with a digitized value being stored for later processing).

DEPR:
If the Crooks et al sequence of FIG. 2 is used for rapid imaging (i.e., where n=1 SE signal for each necessary data signal and no redundant SE signals with different RF phrases are employed for signal averaging), there may be some "ghost" image artifact as depicted in FIG. 2A. However, I have discovered that by reversing the polarity of each successive slice selective G.sub.z gradient on each 180.degree. RF pulse as depicted in FIG. 2B, such image artifacts are eliminated. These artifacts appear as a rotated low intensity duplication of the object that is superimposed on the image object as depicted in FIG. 2A. Such artifacts may be caused by some regions of nuclei achieving only a 90.degree. flip instead of 180.degree. flip (e.g., those located at the edge of a section profile). Any such nuclei also produce signal in the second and later echo image data. In other words, 90.degree.-.tau.-180.degree.-2.tau.-180.degree. sequence becomes, for these regions of nuclei, a 90.degree.-.tau.-90.degree.-2.tau.-90.degree.-.tau. sequence which is analogous to 90.degree.-2.tau.-180.degree.-2.tau. sequence, producing a "stimulated" echo in addition to the desired "Hahn" spin echo. The alternately reversed polarity G.sub.z gradient pulses eliminates these stimulated echoes.

DEPR:
(In performing multi-section imaging with alternating gradient pulse polarities, the RF frequency offsets "hopping" must also have reversed polarity on the negative gradient pulses.) It will be understood throughout the remainder of this discussion, that such alternating G.sub.z polarity gradient pulses may be used, if desired, to eliminate "ghost" image artifact.

DEPR:
Referring now to FIG. 3, it will be seen that to obtain the same amount of data so as to permit a final NMR image having M.times.M pixels, a given extended NMR measurement cycle need be repeated only M/N times using the novel data acquisition procedure of this invention. First of all, it should be observed that, except for the fact that the length of a given cycle is extended, the NMR RF excitation and z-axis gradient pulses and x-axis gradient pulses are substantially the same as in FIG. 2 (as modified by FIG. 2B, if desired). However, it will now be observed that the y-axis magnetic gradient pulses are considerably different. In particular, between each pair of spin echoes within a train of plural spin echoes on a given cycle, there are incremental .delta..phi. y-axis magnetic gradient pulses. Although every other one appears to be of reverse polarity, in actuality they produce cumulative phase effects in the NMR nuclei due to the effect of the interleaved 180.degree. RF nutation pulses. As will be apparent to those skilled in the art, every other spin echo signal is inherently of inverted phase due to the effects of the 180.degree. RF nutation pulse just preceding it. Therefore, to compensate for such effects of the 180.degree. RF nutation pulses used to elicit the spin echo in the first place, the phase of every other spin echo signal must be inverted at some point during signal processing. It is hereinafter assumed, for example, that the phase of all even numbered spin echoes is reversed by simple sign multiplication during digital signal processing.

DEPR:

It will also be observed in FIG. 3 that the cycle dependent y-axis magnetic gradient pulses (e.g. of width .beta.) are incremented by an amount .DELTA..phi. equal to N.times..delta..phi.. To facilitate an understanding of the phase encoding process which occurs on successive spin echo signals elicited during each extended measurement cycle as well as from one measurement cycle to the next, the accumulated spin echo phase magnitude has been line diagrammed for an assumed seven cycles in the lower part of FIG. 3. Thus, starting with the first spin echo in cycle number 1, the relative y-axis phase encoding would be of magnitude +3.DELTA..phi.. The second spin echo signal would have added to that an increment .delta..phi.--as would each subsequent spin echo of that first cycle. It should be noted that because the first incremental spin echo dependent gradient pulse does not occur until after the first spin echo, there are only n-1 such incremental pulses distributed throughout a given measurement cycle. The Nth .delta..phi. increment of phase encoding is effectively added by the cycle-dependent gradient pulse which is incremented by a factor of N.delta..phi.. (In actual practice, to maintain a more symmtric pulse sequence and thus facilitate realization of the pulse train, an extra .delta..phi. pulse may be inserted prior to the first spin echo and the cycle dependent .beta. pulse may be correspondingly offset by 2.delta..phi. so as to compensate threfore.)

DEPR:
After the first measurement cycle has been completed, it can be seen that N spin echo signals have been accumulated, each having successively greater y-axis phase encoding than the last. After a suitable T1 relaxation time, the second cycle will begin with another 90.degree. RF nutation pulse followed by a train of 180.degree. RF nutation pulses. However, on this second cycle, the cycle-dependent y-axis magnetic gradient pulse only has a magnitude of 2.DELTA..phi.. Accordingly, the first spin echo of the second cycle will have only this magnitude of y-axis phase encoding while each subsequent spin echo captured during the second cycle will have an incrementally .delta..phi. greater degree of y-axis phase encoding as should now been apparent and as is depicted in FIG. 3.

DEPR:
To remove the T2-dependency in the collected spin echo data, one must somehow obtain a measure of T2. While various approximate corrections could be devised (e.g. using average T2 factors or the like), it is preferred to obtain sufficient data to actually calculate T2 for each image pixel (--or at least for the neighborhood about each pixel) so that a more accurate compensation for the T2 artifact may be realized. There are probably an infinite variety of techniques that could be utilized for obtaining the requisite T2 data with which to make the compensation. However, one technique is to employ a spatial y-axis gradient pulse sequence so as to produce two sets of identical cycle dependent y-axis phase encoded spin echo data.

DEPR:
Another technique for achieving sufficient data to calculate T2 and thus make T2 corrections is depicted at FIG. 6. Here, relatively larger spin echo-dependent y-axis gradient pulses .delta. are used and relatively smaller cycle dependent y-axis gradient pulses N.DELTA.=.delta. are utilized (this time starting with zero or some initial offset of cycle-dependent phase encoding for the first cycle and then progressively increasing for each successive cycle). Accordingly, after a first complete data acquisition procedure of M/2N cycles, one may compile at TE.sub.1 a sequence of spin echo signals having progressive y-axis phase encoding which starts from "zero" and works up to a magnitude which corresponds to the maximum degree of cycle-dependent phase encoding. Then, in an second data acquisition sequence of M/2N cycles, an extra y-axis magnetic gradient pulse is inserted before the first spin echo thus producing a second sequence of spin echoes occurring at TE.sub.1 but starting from an increment of (M/2N) .delta..phi.. Similarly, the spin echo signals compiled from TE.sub.2 are shifted so as to occupy positions in the y-axis phase encoding plane equivalent to those occupied at TE.sub.1 in the first data acquisition procedure.

DEPR:
Another method of obtaining signals of different .phi. is shown in the sequence diagram of FIG. 3A, which differs from FIG. 3 in that the spin echo is divided into two parts for additional phase encoding. In general, the second half of the echo has a different .phi. than its complementary first half. This .DELTA..phi. is achieved with magnetic gradient pulses G2y and G4y. However, it is equally

important to apply what is termed a "negative time pulse", G3x, at some temporal point between G2x and G4x. In effect, G3x, reverses the focusing of the individual spins so that they are defocused to an earlier time than that of the peak signal. This defocusing permits the G4x pulse to be applied for gradient settling, and then for G5x to be applied during the signal readout period which resamples the peak of the echo. Without the negative time pulse, the signal will have focused beyond the peak at the beginning of G5x or the second readout period (as depicted in FIG. 3B).

DEPR:
Thus, to recap, in the FIG. 3 procedure, a small phase increment, .delta. is applied on each of n-1 echoes in the same train, and a larger phase increment equal of N.delta. is applied to each new cycle of the complete data acquisition sequence. A multiple spin echo pulse train is produced with a 90.degree. RF pulse to produce an FID signal. At time (2n-1) .tau. later 180.degree. RF pulses are applied to refocus the FID signal at times (2n) .tau. and thus to produce spin echo signals. Slice selective G.sub.z gradient pulses are applied on the z-axis of a three Cartesian coordinate system. On a second orthogonal x-axis, a readout G.sub.x gradient is applied at each at each time of spin echo TE to produce spatial encoding of the Fourier transformed signal on the same x-axis. On the third independent Cartesian y-axis multiple pulsed gradients are applied, as shown, at some temporal point before each echo subsequent to the first. All spin echo-dependent gradient pulses have the same strength and pulse duration.

DEPR:
The G.sub.y gradient pulses which encode the even numbered echoes have opposite polarity compared to the gradients on odd numbered echoes. Although the net magnitude of phase accumulated at each spin echo (SE) is linearly increased at each consecutive echo, the sign of the even echo phase change is always opposite the sign of the odd echo phase change due to the phase reversal effect of the intervening 180.degree. RF pulse. By computationally reversing the phase of either the odd numbered echoes or the even numbered echoes, the accumulated y-axis phase encoding increases linearly.

DEPR:
Any number of echoes, N can be acquired with linear phase encoding gradient pulses, however, the T2 decay of later echoes occurring at later echo times TE ultimately limits the number of useful echoes. To acquire all M phase projections needed for a specified M.times.M spatial resolution and image field of view, the pulse train is repeated M/N times. On each additional cycle of the complete acquisition sequence a cycle-dependent G.sub.y gradient pulse before the first spin echo signal is changed by increment of N.delta. where .delta. is the phase accumulated during each of the consecutive echo-dependent gradient pulses. By not acquiring all the phase encoded signals in a single cycle, longer gradient pulses can be used in the same time and the need for extremely fast gradient switching is eliminated. Consequently, larger gradient coils, useful for whole body imaging, can be used for this fast spin echo imaging (the even faster echo planar imaging of Mansfield et al necessitates smaller gradient coils to reduce the effects of coupling to the magnet immediately surrounding the coils).

DEPR:
One potential problem with the present fast spin echo image results from the fact that T2 decay modulates the amplitude of different phase encoded echoes. Thus, different frequencies are weighted by T2 decay and the image is modulated by the reoccurring T2 exponential decay curve (see T2 artifact as shown in FIG. 7).

DEPR:
One process of correcting the data to a common T2 value is shown in FIG. 6. The cycle dependent changes in phase permit the normalization of signals with different amounts of T2 decay, since there are M/N measurements with the same value of T2 decay. For this purpose, a larger increment of echo-dependent phase encoding G.sub.y gradient is used between each echo pair and a smaller increment of G.sub.y phase change is used on each successive cycle. Assuming that the spin-spin relation obeys a monoexponential decay curve, plural measurements at two or more TE can be used to determine the T2 parameter at different spatial locations. This process requires one second set of equivalent phase encoded data to be acquired (e.g. with the "centered" zero phase encoded signals) at a different TE than the first echo. Using only the cycles occurring at the first echoes, 2D-FT produces a 2D spatial image with the y-axis resolution (i.e. the

phase encoded dimension) equal to 1/N the resolution of an image reconstructed from all M echoes at N different times. An image is made with the nth set of data, but using only the nth echo of each cycle for data. Using the two resulting low resolution images, the T2 parameter at each spatial element can be calculated with a mono-exponential fitted to the two magnitude measurements at the same pixel location.

DEPR:
Consider a spatial distribution of nuclei I=1/2 in a magnetic field H.sub.o in the Z-direction on which is imposed a linear gradient G Gauss.multidot.cm.sup.-1 to produce a field dependence on position. The mean precession frequency of the nuclei in absence of gradient is .omega..sub.o =-.gamma.H.sub.o where .gamma. is the gyromagnetic constant. During the period of pulsed gradient applied along the y-axis, the procession phase angle of the nuclei is increased, ##EQU1## where the precession phase angle, .phi., is defined in a reference frame rotating at .omega..sub.o. As shown in the gradient pulse sequence diagrams, G.sub.y pulses are applied every 2.tau. interval. The 180.degree. RF pulses nutate the spins and change the sense of the net phase existing prior to the pulse. If there is a phase shift of .DELTA..phi. before the second 180.degree. RF pulse, then immediately following the nutation process, .DELTA..phi.(4.tau.)=.DELTA..phi.(3.tau.). To encode a constant net accumulation of phase shifts at successive echoes, the polarity of alternate gradient pulses is changed, so that ##EQU2## where n denotes the nth successive echo in the train. In these experiments, the total number of echoes in the train, N.sub.E, is limited to 8 with a 14 msec interval due to the time required for the switching gradient pulses (msec) and for radio frequency (RF) irradiation. The RF pulses are modulated by a truncated sinc function which acting with a slice selective gradient, G.sub.z, has a frequency spectrum matching the resonant frequencies of nuclei located within a planar volume. The pulse sequence was iteratively repeated 16 times with different incremental gradient strengths G.sup.1.sub.y during cycle-dependent pulse duration, .beta., to change the phase at all n echoes ##EQU3## where c is the cycle number of the pulse sequence iteration of total cycle number N.sub.c. The phase interval of each cycle equals a fraction of the phase increment accumulated by each consecutive echo, G.sub.y.sup.1 =G.sub.y.sup.2 /N.sub.c so that the total set of the N.sub.c .multidot.N.sub.E echoes has a single uniform increment of phase difference. In practice, the temporally consecutive echoes are regrouped to provide a data set having a linear increase in phase before applying a Fourier transform. Also note that spatial resolution on the x-axis is produced by the presence of G.sub.x during the spin echo formation time (centered at n.multidot.2.tau.), which imposes a linear dependence between the resonance frequency of nuclei and their position on the x-axis. The image, which is the density distribution of nuclei, .rho. (x,y), is reconstructed from these spin echo signals, M(t), by 2D-Fourier transformation, ##EQU4## where H(.phi.)=e.sup.-nTE/T2 and TE is the time of the echo formation. The modules or magnitude of .rho.(x,y) is displayed as the actual intensity of the digital image. The spin echo signals are detected in quadrature for determination of phase, and converted to digital data for computer image processing.

DEPR:
A novel method of removing the horizontal line artifacts and other T2 decay effect from the image is especially applicable since the data is not acquired at a single TE. Two of these images with different TE can be used to produce a spatial map of T2 which is thereafter used to normalize all data to the same effective TE. A modification of the pulse sequence may then be used to produce two sets of identical cycle dependent phase encoded echoes with phase centered on an assumed .phi.=0. For example, by centering the cycle dependent gradient pulses on G.sup.1.sub.y =0, removal of .delta.G.sup.1.sub.y immediately prior to the second 180.degree. RF pulse, and placing a pulse of area -4.delta.G.sup.1.sub.y immediately prior to the second 180.degree. RF pulse, two sets of identically phase encoded echoes are defined at times 2.tau. and 6.tau.. The acquisition of an additional echo is required to maintain the maximal accumulated net phase and corresponding spatial resolution. ##EQU5## Spatial resolution is reduced by a factor of 1/N.sub.E on the phase encoded dimension in these images since the net phase does not accumulate on successive echoes. The image field of view remains inversely proportional to the cycle dependent phase increment .beta.G.sub.1.sup.y chosen to prevent aliasing. Consequently, the T2 map is made with T2 averaged in the phase encoded axis, with no averaging of spatial resolution on the x-axis encoded independently by G.sub.x T2(x,y) is used to numerically scale the

similarly 2-D Fourier transformed data taken at other single echo times, so that the moduli of the complex numbers in each pixel are extrapolated to the same echo time (the choice of TE is arbitrary). Each such scaled complex z data set is inverse Fourier transformed and the T2 normalized time domain data is reshuffled into a single data set which is N.sub.E times larger. A final forward Fourier transform of the T2 normalized data produces the image .rho.(x,y) deconvolved from H(.phi.). It should be noted that although the image dependence on T 1 is determined by the TR of data acquisition, the T2 map can be used to scale data to several different TE values for synthesis of multiple spin echo images which have known clinical utility.

CLPV:
eliciting and recording plural NMR spin echoes from a selected slice of the object while encoding a first dimension of spatial information therein by imposing a magnetic field gradient G.sub.x along an x-axis dimension during readout of the recorded spin echo signals;

CLPV:
imposing a cycle-dependent magnitude of magnetic field gradient G.sub.y along a y-axis dimension at some time during each cycle to achieve spatial phase encoding along said second dimension; and

CLPV:
imposing a spin-echo-dependent magnitude of magnetic field gradient G.sub.y along said second dimension prior to the occurrence of at least some spin echo signals in a given cycle so as to achieve further degrees of y-axis phase encoding within different spin echo signals in a given cycle.

CLPV:
(b) imposing a first incremental phase change .delta..phi. in at least some said spin echoes by applying at least one magnetic gradient pulse directed along said predetermined coordinate axis between occurrences of at least some of said spin echoes;

CLPV:
(c) imposing a secord incremental phase change n.DELTA..phi. in at least some of said spin echoes by applying at least one additional magnetic gradient pulse directed along said predetermined coordinate axis at least once during said eliciting step (a);

CLPV:
means for eliciting and recording plural NMR spin echoes from a selected slice of the object while encoding a first dimension of spatial information therein by imposing a magnetic field gradient G.sub.x along an x-axis dimension during readout of the recorded spin echo signals, said plural spin echoes being elicited in each of plural acquisition cycles, each of which cycles also elicits a train of plural spin echoes by exciting nuclei within said selected slice with a first NMR rf signal pulse followed by a train of 180.degree. NMR rf signal pulses;

CLPV:
means for imposing a cycle-dependent magnitude of magnetic field gradient G.sub.y along a y-axis dimension at some time during each cycle to achieve spatial phase encoding along said second dimension; and

CLPV:
means for imposing a spin-echo-dependent magnitude of magnetic field gradient G.sub.y along said second dimension prior to the occurrence of at least some spin echo signals in a given cycle so as to achieve further degrees of y-axis phase encoding within different spin echo signals in a given cycle.

CLPV:
imposing a spatially selective magnetic gradient pulse G.sub.z upon said object during the occurrence of each said rf nutation pulse so that only nuclei in a selected portion of said object are thereby nutated;

CLPV:
the polarity of said G.sub.z magnetic gradient pulse periodically having opposite values for different ones of said nutation pulses; and

CLPV:
imposing G.sub.x and G.sub.y phase encoding magnetic gradient pulses and recording NMR spin echo data resulting from said 180.degree. rf NMR nutation pulses.

CLPV:
dividing the NMR spin echo response into separate differently phased portions by applying to said object a magnetic gradient pulse of a first polarity between a pair of opposite polarity magnetic gradient pulses during the occurrence time of the spin echo response.